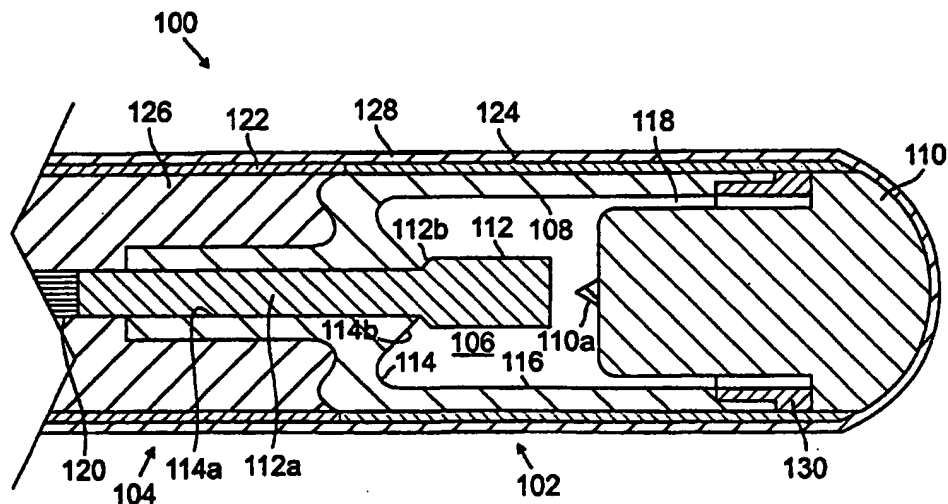




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(54) Title: X-RAY CATHETER



(57) Abstract

A catheter for emitting radiation is disclosed, comprising a catheter shaft (104), and an x-ray unit (102) attached to the distal end of the catheter shaft. The x-ray unit comprises an anode (112), and a cathode (110) coupled to an insulator (108) to define a vacuum chamber (106). The cathode is preferably a field emission cathode of graphite or graphite coated with titanium carbide, for example. The anode is preferably tungsten, and the insulator is preferably pyrolytic boron nitride. The x-ray unit is preferably coupled to a voltage source through a coaxial cable. The anode is preferably a heavy metal such as tungsten. The cathode may also be a ferroelectric material. The x-ray unit can have a diameter less than about 4mm, and a length less than about 15 mm. Methods of use of the catheter are also disclosed. The catheter of the present invention can be used to irradiate the site of an angioplasty procedure to prevent restenosis. It can also be used to treat other conditions in any vessel, lumen or cavity of the body.

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1

X-RAY CATHETER

This application claims the benefit of U.S.
5 Provisional Application Nos. 60/006,708 filed November 14,
1995, and 60/002,722 filed August 24, 1995.

FIELD OF THE INVENTION

The present invention relates generally to catheters
and, more particularly, to catheters for irradiating
10 vessels, lumens or cavities of a body, such as
cardiovascular tissue to reduce the incidence of restenosis,
and to treat other conditions.

BACKGROUND OF THE INVENTION

Restenosis of an artery or vein after percutaneous
15 transluminal coronary angioplasty (PTCA) or percutaneous
transluminal angioplasty (PTA) occurs in about one-third of
the procedures, requiring the procedure to be repeated.
Various types of drugs or other agents are being
investigated for use in preventing restenosis. Heparin, an
20 anticoagulant and inhibitor of arterial smooth muscle
proliferation, is one such drug. Dexamethasone may also
prevent smooth muscle proliferation. Integralin, which
prevents aggregation of platelets, may also be useful.
Other anticoagulants and antiproliferative agents are being
25 investigated for efficacy, as well. Such drugs can be
delivered before or after the angioplasty procedure. The
delivery of lytic agents such as urokinase, streptokinase
and recombinant tissue type plasminogen activator (rTPA) to
dissolve thrombi in arteries and veins is also being
30 investigated.

Because of blood flow through the artery, drugs
delivered to the site of an angioplasty procedure, for
example, can be rapidly dissipated and removed from the site
before they can be sufficiently absorbed to be effective.
35 Catheters have therefore been developed to directly drive
the drug into the desired site through a balloon or to
maintain the delivered drug agent proximate the desired site
by isolating the region with occlusion balloons. See, for

1 example, U.S. Patent Nos. 5,087,244, 4,824,436, and
4,636,195, to Wolinsky.

5 The use of sufficient pressures to drive the drug
into the tissue or plaque, however, may damage the arterial
wall. Passive delivery into a region isolated by occlusions
balloons, on the other hand, is slow and may not enable
sufficient absorption of the medication. Passive delivery
can be particularly inappropriate for drug delivery in an
artery because blood flow can only be occluded in an artery
10 for a limited period of time.

 Stents have also been used after angioplasty to
prevent an opened blood vessel from closing. The use of
stents, however, has only shown a small decrease in the
incidence of restenosis. Stents are also difficult to
15 properly position and are expensive.

 The use of radiation has also been investigated to
reduce restenosis after PTCA or PTA. One technique is
Photodynamic Therapy (PDT), wherein photosensitive drugs
delivered to the angioplasty site are activated by
20 irradiation with ultraviolet (UV) or visible light.

 Another approach was to expose vascular tissue to UV
light within a wavelength band of DNA absorption (240-280
nm) by a laser to disable or destroy the DNA of the tissue.
This would impair or destroy the ability of the vascular
25 tissue to proliferate. This approach had only limited
success, however, because UV light does not penetrate
vascular tissue sufficiently to prevent proliferation or
migration of smooth muscle tissue.

 Beta-irradiation of the vessel after angioplasty
30 with radioactive guide wires or implanted stents is another
technique. U.S. Patent No. 5,199,939 to Dake et al., for
example, discloses a catheter with radioactive pellets at
its distal end to irradiate the site of an angioplasty
procedure to prevent restenosis. The need for a radioactive
35 source in the catheter lab, however, requires protection
against radioactive hazards to personnel and costly
compliance with regulations. It is also difficult to
control the depth of penetration of the radiation by this
method.

1 U.S. Patent No. 4,143,275 to Mallozzi et al.,
discloses an x-ray device for delivering radiation to remote
locations of the human body such as the interior of the
heart. The x-ray radiation is generated by irradiating a
5 target material, such as iron, calcium, chromium, nickel,
aluminum, lead, tungsten or gold, by a laser to vaporize the
metal. X-ray radiation is emitted from the ionized vapor
plasma. The target is located outside the body and the x-
rays are directed to a desired location within the body
10 through a hollow guide. The patent discusses use of such a
device to produce radiographs, to irradiate tumors or to
alter tissue. It is believed, however, that x-ray radiation
generated by this method would have photon energy of about
1-2 KeV at best, which is too low to penetrate biological
15 tissue deeper than about 20-30 microns. In addition, the
patent does not disclose how to produce a guide which is
both flexible enough to be advanced through the
cardiovascular system and able to transmit adequate x-ray
radiation to an intended site without excessive losses.

20 U.S. Patent No. 5,153,900 to Nomikos, et al.,
discloses a miniaturized low power x-ray source for
interstitial insertion for the treatment of tumors. The
device comprises a housing with an elongated cylindrical,
rigid probe. An anode and cathode are located in the
25 housing and a target is located at the distal end of the
probe. The cathode and target must lie along the same axis.
Electrons emitted by the cathode, which can be a thermionic
emitter or a photocathode, impinge on the target, causing
the emission of x-ray radiation. A rigid probe is
30 unsuitable for use in the cardiovascular system.

U.S. Patent No. 5,428,658 to Oettinger, et al., a
continuation of the patent to Nomikos, discussed above,
discloses a flexible probe comprising a flexible optical
fiber within a metallic tube. The optical fiber has a
35 photoemissive coating at its terminal end. A target is
located distal to the terminal end of the optical fiber,
within an evacuated shell. The flexible probe is said to
enable threading down a pathway, such as the trachea, or
around structures, such as nerves or blood vessels. Such a

1 device is not sufficiently flexible for advancement through
the cardiovascular system, nor is it believed that such a
device can be made small enough to access the site of a PTCA
procedure.

5 U.S. Patent No. Re 34,421 to Parker, et al.
discloses an x-ray microtube comprising a glass tube having
a diameter less than one inch, for insertion into the body
for treating a tumor. While asserting that the diameter can
be as small as 1/8 inch, Parker does not address any of the
10 problems associated with such a small device, such as
electrical flashover. It is questionable whether such a
device could be made small enough to access the site of a
PCT procedure, and still function. Glass also has too high
a coefficient of absorption of x-ray radiation to enable
15 delivery of sufficient x-ray radiation to prevent restenosis
in a reasonable period of time. Parker also does not
disclose any way to advance its x-ray source through the
cardiovascular system, or any other channel of the body.

SUMMARY OF THE INVENTION

20 In accordance with a preferred embodiment of the
present invention, an x-ray catheter is disclosed which is
small and flexible enough to access an intended site within
a vascular system of the body, such as the coronary arteries
of the cardiovascular system. The x-ray catheter can
25 operate at the high voltages required for generating x-ray
radiation of an effective spectrum for preventing restenosis
and treating other conditions. It also has walls highly
transmissive to x-ray radiation so that an effective dosage
can be delivered in a short period of time.

30 In accordance with the present invention, a catheter
for emitting x-ray radiation is disclosed comprising a
flexible catheter shaft having a distal end and an x-ray
unit coupled to the distal end. The x-ray unit comprises an
anode, a cathode and an insulator, wherein the anode and
35 cathode are coupled to the insulator to define a vacuum
chamber. The insulator is preferably pyrolytic boron
nitride, which is highly transmissive to x-ray radiation.
The cathode is preferably a field emission cathode of
graphite, graphite coated with titanium carbide, or other

1 carbides. The cathode can also comprise silicon and the x-ray unit can include a grid. The cathode can be a ferroelectric material, as well. The anode is preferably tungsten. The catheter shaft is preferably a coaxial cable.
5 A guide wire may be provided extending through the catheter shaft, partially through the catheter shaft or partially through the x-ray unit, in a rapid exchange configuration. The catheter further preferably comprises a means for centering the x-ray unit within a lumen.

10 In accordance with another embodiment of the invention, an x-ray catheter is disclosed comprising a flexible catheter shaft for being advanced through lumens of a vascular system.

Another embodiment of the present invention
15 comprises an x-ray generating unit having a diameter less than about 4 mm.

Yet another embodiment of the present invention comprises a catheter shaft, an x-ray generating unit and means for centering the x-ray generating unit within the
20 lumen.

A method is also disclosed in accordance with the present invention for preventing restenosis of a lumen or treating other conditions, comprising advancing an x-ray catheter through a lumen to a first location adjacent an
25 intended site of the lumen, wherein the x-ray catheter comprises a flexible catheter shaft with a distal end and an x-ray generating unit coupled to the distal end. The x-ray generating unit comprises an anode, a cathode and an insulator, wherein the anode and cathode are coupled to the
30 insulator to define a vacuum chamber. The method further comprises causing the emission of an effective dose of x-ray radiation and removing the catheter. The catheter can be inserted after conducting an angioplasty procedure. The catheter can be advanced over a guide wire and through a
35 guide catheter, or through an exchange tube.

DESCRIPTION OF THE FIGURES

Fig. 1A is a cross-sectional view of an x-ray catheter in accordance with a first embodiment of the present invention;

- 1 Fig. 1B is a cross-sectional view of a preferred catheter shaft for use in the present invention;
- Fig. 2A is a graph of an exemplary voltage applied between the anode and grid electrode versus time;
- 5 Fig. 2B is a graph of an exemplary voltage applied between the grid electrode and rear electrode of the cathode versus time;
- Fig. 2C is a graph of the current flow from the cathode to the anode versus time, for the voltages of Figs.
- 10 2A and 2B;
- Fig. 2D is a graph of the power of the emitted x-ray radiation for the voltages of Figs. 2A and 2B;
- Fig. 3A is an alternative cathode in accordance with a second embodiment of the invention;
- 15 Fig. 3B is an enlarged cross-section of one needle of Fig. 3A;
- Fig. 4 is a graph of photon energy versus the Linear Attenuation Coefficient, μ ;
- Fig. 5 is a cross-sectional view of the distal
- 20 portion of a third embodiment of the present invention;
- Fig. 6 is a cross-sectional view of mandrel for use in chemical vapor deposition of the insulator of the embodiment of Fig. 5;
- Fig. 7 is a cross-sectional view of the distal
- 25 portion of a fourth embodiment of the present invention;
- Fig. 8 is a cross-sectional view of the distal portion of a fifth embodiment of the present invention;
- Figs. 9-11 are side views of the distal portions of the catheter of the present invention, including several
- 30 centering devices for centering the x-ray unit within a lumen;
- Fig. 14 is a cross-sectional view of a distal portion of a catheter in accordance with the present invention, in a rapid exchange configuration wherein the
- 35 guide wire passes through the distal tip of the x-ray unit; and
- Fig. 15 is a partial cross-sectional view of another catheter in accordance with the present invention in a rapid exchange configuration wherein the guide wire enters and

1 exits the catheter shaft proximal to the x-ray unit.

DESCRIPTION OF THE INVENTION

5 Fig. 1A is a cross-sectional view of an x-ray catheter 10 in accordance with a first embodiment of the present invention. The x-ray catheter 10 comprises a flexible catheter shaft 12 adapted for insertion into blood vessels or other body vessels. The shaft 12 can be
10 polyethylene, polyurethane, polyether block amide, nylon 12, polyamide, polyamide copolymer, polypropylene, polyester copolymer, polyvinyl difluoride or silicon rubber, for example.

A miniature x-ray unit 14 is secured at the distal end of the catheter shaft 12 by an adhesive, for example.
15 The x-ray unit 14 comprises a vacuum chamber 16, a cathode 18, which emits electrons, and an anode 20, which receives the emitted electrons. The anode 20 abruptly decelerates the impinging electrons, causing the emission of x-ray radiation by the Bremsstrahlung effect, as is known in the
20 art. About 0.1-0.2% of the kinetic energy of the impinging electrons is emitted in the x-ray range of about 0.5-5 Angstroms in the preferred embodiments of the present invention.

In this embodiment, the anode 20 preferably has the
25 shape of an inverted cone. The walls of the anode 20 preferably have an angle of about 16° with respect to the surface of the cathode 18. The anode 20 is preferably a heavy metal, such as gold or tungsten, for example.

The cathode 18 comprises a base 19 which in this
30 embodiment is preferably a ferroelectric material, as discussed below. The base 19 can also be doped or undoped silicon, or other such materials, which is also discussed below.

A grid electrode 24 is coupled to the surface of the
35 base 19 facing the anode 20. A rear electrode 27 is coupled to the rear of the base 19. Wires 26, 28 and 30 extend from the rear electrode 27, anode 20 and the grid 24, respectively, through the catheter shaft 12, to a high voltage generator 32. The generator 32 preferably operates

1 in the 0-30 kilovolt (Kv) range. The wires 26, 28 and 30
can be soldered in place. Separate lumens 34, 36, 38 can be
provided through the catheter shaft 12 for each wire or a
single lumen can be provided for a coaxial cable comprising
5 the three wires. A coaxial cable can form the catheter
shaft as well, as in the embodiments of Figs. 5 and 7.

The vacuum chamber 16 preferably comprises a wall 22
of beryllium, beryllium oxide, aluminum, aluminum oxide,
pyrolytic boron nitride, graphite or other such metal or
10 ceramic materials, which is transparent to x-rays. If a
metal, such as beryllium or aluminum is used as the wall 22
of the vacuum chamber 16, an insulative layer (not shown)
would be provided to electrically insulate the anode 20 and
cathode 18, as is known in the art. Aluminum oxide,
15 pyrolytic boron nitride and other ceramics are insulators.
A transparent biocompatible coating 25 of a polymeric
material such as polyethylene, polyurethane or Teflon (R),
for example, is also provided over the wall 22. A vacuum
tie off (not shown) depends from the vacuum chamber 16,
20 which is sealed after the desired vacuum within the chamber
is achieved. A soft, resilient material 48 may be provided
at the distal tip of the x-ray unit 14, as is known in the
art. The material can be ultra low density polyethylene or
nylon, for example.

25 A lumen 40 extending longitudinally through the
catheter shaft 12 can also be provided to accommodate a
guide wire 42. A port 44 can be provided through the shaft
12 for the guide wire 42 to exit the shaft 12. A tube 48
can be attached by adhesive or thermal bonding to the shaft
30 12 at the port 44 to provide a guide for the guide wire 42
around the x-ray unit 14. The tube 48 may be adhered to
the wall of the x-ray unit 14, as well. The tube 48 may
extend through the soft material 46 at the distal tip of the
x-ray unit 14.

35 The lumens in Fig. 1 are shown in the same plane for
illustrative purposes. If multiple lumens are provided,
they would preferably be arranged symmetrically within the
catheter, as shown in Fig. 1B.

1 In this embodiment, the base 19 of the cathode 18 is
preferably a ferroelectric material, as described in Riege,
H., "Electron emission from ferroelectrics - a review,"
Nuclear Instruments and Methods in Physics Research A340
5 (1994), pp. 80-89; Gundel, H., et al., "Fast Polarization
Changes in Ferroelectrics and Their Application," Nuclear
Instruments and Methods in Physics Research A280 (1989), pp.
1-6; Gundel, H., et al., "Time-dependent electron emission
from ferroelectrics by external pulsed electric fields," J.
10 Appl. Phys. 69(2) 15 January 1991, pp. 975-982; and Asano,
Jun-ichi, et al., "Field-Excited Electron Emission from
Ferroelectric Ceramic in Vacuum," Jpn. J. Appl. Phys. Vol.
31 (1992), pp. 3098-3101, Part 1, No. 9B, which are all
incorporated by reference herein. As described in those
15 articles, ferroelectric materials, such as lead-zirconium-
titanate (PZT) and lead-lanthanum-zirconium-titanate (PLZT)
and triglycinesulfate (TGS), for example, emit electrons
from their surfaces when the spontaneous ferroelectric
polarization of these materials is rapidly reversed. High
20 voltage, submicrosecond pulses can cause such reversals, as
can mechanical pressure pulses, thermal heating or laser
illumination. The use of a laser to cause polarization
reversal is discussed in Geissler, K., et al., "Intense
laser-induced self-emission of electrons from
25 ferroelectrics," Physics Letters A 176 (1993), pp. 387-392,
North Holland, which is also incorporated by reference
herein. Ferroelectric cathodes do not require as high
vacuum as other types of cathodes. A vacuum of about 10^{-3} -
 10^{-4} Torr is sufficient. Ferroelectric cathodes are also
30 simple to manufacture and are reliable.

 Preferably, the polarization switching is caused by
applying an electrical pulse across the ferroelectric
material. Preferably, voltage pulses are applied between the
rear electrode 27 and the grid electrode 24. Positive or
35 negative pulses, or a combination of positive and negative
pulses, can be used, depending on the configuration and
original orientation of the polarization of the
ferroelectric material. The reversal of ferroelectric
polarization can be achieved by applying a voltage pulse of

1 between about 1-3 Kv to the ferroelectric cathode 18 via the
rear electrode 27 and the grid electrode 24. The pulses are
preferably applied for 5-100 nanoseconds. The polarization
of the ferroelectric material 19 can be switched at a rate
5 of between about 1 kHz-5 MHz. Electrical current densities
as high as 100 Amps per square centimeter can be generated.
With a polarization switching rate of about 100 kHz, for
example, and a diameter of ferroelectric material 19 of
about 1 mm, an average anode current of about 10
10 milliamperes can be generated.

Preferably, a constant voltage or voltage pulses are
applied between the anode and the cathode, as well, to
control the energy of the emitted x-ray radiation, and hence
the depth of penetration of the radiation into tissue. A
15 voltage of about 10-30 Kv is preferred in coronary
applications, as discussed further, below.

In this embodiment, the grid electrode 24 is
preferably silver, aluminum or gold. About one-half of its
area is transparent or open to electrons. The grid 24 can
20 be deposited on a layer of ferroelectric material, such as
PZT, PLZT or TGS, as is known in the art. The dimensions of
the cathode 18 depend on the application. For use in
coronary arteries, for example, the ferroelectric material
19 can have a diameter of about 1-2 mm. For use in larger
25 blood vessels, such as the femoral artery, the diameter of
the ferroelectric material 19 could be up to 3 mm. The
thickness of the ferroelectric material 19 can be between
about 50-1,000 microns. About 200-500 microns is preferred.
The grid 24 is preferably about 0.5-10 microns thick, with
30 about the same diameter as the ferroelectric material 19.
The electrode 27 is about 1 micron thick. The distance
between the anode 20 and cathode can be about 0.2-5 mm.

Experimental data suggests that restenosis after
PTCA can be limited by irradiation by about 2000 centigrays
35 (cGy). (See, for example, Tim A. Fischel et al., "Low-Dose,
beta-particle emission from "stent" wire results in
complete, localized inhibition of smooth muscle cell
proliferation," Circulation, Vol. 90, No. 6, December 1994,
and Wiedermann, Joseph G., et al., "Intracoronary

1 Irradiation Markedly Reduces Neointimal Proliferation After
Balloon Angioplasts in Swine: Persistent Benefit at 6-Month
Follow-Up," JACC Vol. 25, No. 6, May 1995, 1451-6, which are
incorporated by reference, herein).

5 It is believed that the x-ray unit in accordance
with this and the other embodiments of the present invention
disclosed herein can emit over 2000 centigrays of x-ray
radiation in about one minute, to a cylindrical region of a
lumen with a length of about 5 mm. Treatment of a typical
10 lesion in a coronary artery, which can be 1-2 centimeters
long, can require repositioning of x-ray unit several times
to irradiate the entire lesion. A lesion 1-2 centimeters
long can therefore be irradiated in about 2-5 minutes. The
x-ray catheter of the present invention can deliver
15 sufficient x-ray radiation to a lesion in a short period of
time which minimizes the inconvenience and discomfort of the
patient and cost of the procedure.

In operation, the high voltage generator 32
preferably applies voltage pulses between the anode 20 and
20 grid 24, and between the rear electrode 27 and grid 24. In
Fig. 2A, exemplary voltage pulses applied between the anode
20 and grid 24, V_{AG} , are plotted versus time. The voltage
pulses in this example are about 10-12 Kv. The voltage
pulses between the anode 20 and grid 24 can be applied for
25 about 0.1-1.0 microseconds, every 10 microseconds. Fig. 2B
plots exemplary voltage pulses V_{GR} , applied between the grid
electrode 24 and the rear electrode 27 versus time. The
voltage difference here is about 2.0 Kv. Fig. 2B also shows
a negative pulse 49 which is preferably applied to restore
30 the negative charge on the surface of the ferroelectric
material 19 adjacent the grid 24. Fig. 2C illustrates
qualitatively the current I_A flowing from the ferroelectric
material 19 to the anode 20 for the voltage pulses shown in
Figs. 2A and 2B. The length of each current pulse generated
35 for the range of voltage pulses of 0.1-1 microsecond, is
about 10-100 nanoseconds. The current pulses cause the
emission of pulses of x-ray radiation with peak power in
this example of up to about 30 watts, as shown in Fig. 2D.

1 In a second embodiment of the invention, shown in
Fig. 3A, the cathode 18 may also be a field emission cathode
50 comprising multiple needles 52 and optionally a grid
electrode 54. Fig. 3B is an enlarged cross-sectional view
5 of a single needle 52, of Fig. 3A. The base 55 and needles
52 can be doped or undoped silicon. The grid 54 can be
niobium. If a grid 54 is provided, a layer 57 of an
insulator, such as silicon dioxide (SiO_2), is preferably
deposited over the base 55 of silicon. The grid 54 of
10 niobium is deposited over the silicon dioxide layer 57. A
rear electrode 59 is coupled to the rear of the base 55. A
wire 58 is coupled to the rear electrode 59. A wire 56 is
coupled to the grid 54. Returning to Fig. 3A, a vacuum tie-
off 60 is shown, as well. The anode 20 can be the same as
15 described above.

The radius of the tips of the needles 52 is between
about 5-100 Angstroms. The height of the needles is about
0.5-1.0 microns. The grid 54, which is about 0.5 microns
thick, is preferably positioned slightly above the top of
20 the needle 52, as shown in Fig. 3B. The openings in the
grid 54 have a diameter of about 2 microns. The layer of
silicon dioxide is about 1-2 microns thick. A vacuum of
between about 10^{-7} - 10^{-8} Torr is preferred for a field
emitting cathode including silicon.

25 The needles 52 emit electrons when negative
potential is applied between the rear electrode 59 and the
grid electrode 54. A triggering voltage of about 100-500
volts may be used, for example. The voltage can be constant
or pulsed. If no grid electrode is provided, the high
30 voltage can be provided directly between the anode and the
needles 52.

The radiation emitted by the anode 18 passes through
the vacuum chamber wall 22 and coating 25, into surrounding
tissue. Irradiation reduces the ability of smooth muscle
35 cell to proliferate, inhibiting restenosis, as discussed
above. Fig. 4 is a graph of Photon Energy (kev) versus the
Linear Attenuation Coefficient μ (cm^{-1}) for bone 62, muscle
64 and lung tissue 66. (See, Anthony Brinton Wolbarst,
Physics of Radiology, Appleton and Lange, 1993, p. 108;

1 Johns, H.E., Cunningham, JR.: The Physics of Radiology, 4th
ed., Springfield, IL; Charles C. Thomas, 1983, Appendix A.)
The greater the coefficient μ , the more effectively the
medium absorbs and scatters photons. The depth of
5 penetration of radiation is the depth at which the intensity
of the impinging radiation drops to $1/e$ of its original
value. The depth of penetration of x-ray radiation of a
particular energy is equal to $1/\mu$. Generally, the
coefficient μ increases with increasing effective atomic
10 number of the material. While muscle and lung tissue have
nearly identical chemical composition, the attenuation in
muscle tissue is about 3 times greater than the attenuation
in lung tissue, because muscle tissue is about 3 times
denser than lung tissue. The energy of x-ray radiation is
15 preferably adjusted so that it penetrates slightly into the
adventitia tissue of the blood vessel about 2 mm deep.
Penetration into the cardiac muscle tissue beyond the
coronary artery, for example, should be minimized. The
energy can be adjusted by varying the voltage applied
20 between the anode and cathode. The preferred voltage range
of 10-30 Kv yields x-ray radiation with a peak energy of
about 8-10 KeV, which is appropriate in coronary
applications.

Fig. 5 is a cross-sectional view of the distal
25 portion of an x-ray catheter 100 in accordance with a third
embodiment of the present invention. The x-ray catheter 100
comprises an x-ray unit 102 coupled to a high voltage
coaxial cable 104. The x-ray unit 102 has a vacuum chamber
106, defined by an insulator 108, a cathode 110 and an anode
30 112. The insulator 108 comprises a base portion 114 coupled
to a tubular, preferably cylindrical wall portion 116 with
an open end 118. The cathode 110, which is a cold, field
emission cathode, is coupled to the open end 108. The
insulator 108 is preferably alumina, beryllium oxide or more
35 preferably, pyrolytic boron nitride. The boron nitride must
be pyrolytic, as opposed to sintered, because only the
pyrolytic boron nitride is vacuum tight at the wall
thicknesses required. The cathode 110 is preferably
graphite. The anode 112 is preferably tungsten or tungsten

1 coated with a layer of platinum. A one micron layer of
platinum is sufficient. The vacuum is preferably 10^{-5} Torr
or better.

5 The cathode 110 is preferably graphite, carbides,
such as titanium carbide, silicone, metals, or graphite
coated with titanium carbide. The cathode 110 preferably
includes one or a plurality of protrusions 110a with a sharp
tip extending towards the anode 112 along a central axis of
the x-ray unit 102. The protrusion 110a locally enhances
10 the electrical field and improves the emission of electrons,
as is known in the art. The protrusion 110a can comprise
the same material as the cathode 110, or can be another of
the cathode materials mentioned above.

The anode 112, which is preferably in the shape of a
15 rod, extends along the central axis of the x-ray unit 102.
The rod 112 has a depending portion 112a received within a
cylindrical groove 114a extending through the base portion
114. Preferably, the base 114 has a portion 114b, which
tapers toward the anode 112. An angle of about 45° can be
20 used, for example. The anode 112 also can have a portion
112b tapered toward the cylindrical portion 114b of the
base. Such a configuration displaces the electrical field
from the anode-vacuum-insulator triple junction, decreasing
the risk of electrical flashover during operation. The
25 anode 112 is preferably a heavy metal. Tungsten is
preferred.

The cathode 110 and anode 114 are coupled to the
high voltage generator 32 of Fig. 1, described above,
through the high voltage coaxial cable 104. The coaxial
30 cable 104 comprises a central conductor 120, which is
coupled to a proximal end of the anode 114, and an external
conductor 122, which is coupled to the cathode 110. A
conductive coating 124 is provided over the external surface
of a portion of the cathode 110 and the external surface of
35 the insulator 108 to couple the cathode 110 to the external
conductor 122. A silver coating with a thickness of about
0.1-1.0 microns may be used. Gold may be used as well.
Insulation 126, such as Teflon (R), silicone, rubber,
fluorinated ethylene propylene (FEP) or polyethylene, for

1 example, is typically provided between the external
conductor 122 and the central conductor 120. The x-ray unit
102 can be attached to the coaxial cable 114 with an
adhesive, for example.

5 The cathode's "triple junction point" (the junction
between the cathode, the insulator and the vacuum), which in
this embodiment is an annular region surrounding the cathode
110 proximate the open end 118 of the insulator 108, is
screened from the high electrical field between the anode
10 112 and the cathode 110 by the conductive coating 124 and
the side of the cathode 110. This decreases the incidence
of electrical flashover, enabling the use of higher
voltages.

 The cathode 110 can be coupled to the open end 118
15 of the insulator 108 through a metal ring 130. The metal
ring can comprise tungsten, platinum, or graphite covered by
platinum. Coupling of the cathode 110 to the metal ring and
coupling of the anode 112 to the insulator 108 is described
further, below.

20 A biocompatible layer 128 is provided over the
external conductor 116, conductive layer 124, and the
cathode 110. A thickness of less than about 0.002 inches is
preferred. Preferably, the biocompatible coating 128 also
acts as an insulating layer. The biocompatible coating may
25 be silicone or FEP, for example. A lubricious layer (not
shown) of a hyaluronic coating, for example, may be provided
as well. The biocompatible coating may have sufficient
lubricity without a further coating. Silicone, for example,
is a highly lubricious biocompatible coating.

30 The coaxial cable 104 is chosen to have sufficient
flexibility to be advanced through the cardiovascular or
other such system, to an intended site. It has been found
that standard high voltage coaxial cables are generally not
flexible enough to be advanced through the cardiovascular
35 system to the coronary arteries. It has further been found,
however, that miniature high frequency coaxial cables are
available with sufficiently small diameter (about 1.0-3.0 mm
outer diameter) and sufficient flexibility to be advanced to
the coronary arteries. Usually, such cables are used in

1 high frequency applications at voltages less than several
kilovolts. Surprisingly, it has been found in connection
with the present invention, that these cables can hold
direct current voltages as high as 75-100 Kv without
5 breakdown, and consequently can be used with the x-ray unit
of the present invention for operational voltages of up to
30-40 Kv. Such voltages are sufficient to generate x-ray
radiation in appropriate energy ranges for the treatment of
restenosis and other conditions. Suitable coaxial cables
10 include CW2040-3050FR; CW2040-30; CW2040-3675-SR; and
CW2040-3275SR, distributed by Cooner Wire, Inc. Chatsworth,
CA, for example. Cooner distributes coaxial cables for New
England Electric Wire Corporation, Lisborn, New Hampshire.

An x-ray unit 102 in accordance with this embodiment
15 of the invention can have a length less than about 15 mm and
a diameter less than about 4.0 mm, depending on the
application. The distance between the cathode 108 and the
anode 110 can be between about 2.0-0.2 mm, depending on the
size of the x-ray unit 102. The thickness of the
20 cylindrical insulator wall 116 can be between about 0.2-0.5
mm. The diameter of the coaxial cable 104 can be about the
same as the diameter of the x-ray unit 102. For use in
preventing restenosis after dilatation of a coronary artery,
which typically has a diameter of about 3 mm, the x-ray unit
25 102 preferably has a length of about 7 mm and a diameter of
about 1.5 mm. In peripheral blood vessels, which are
larger, the x-ray unit 102 preferably has a diameter of
about 3.5 mm and a length of between about 7-15 mm. Larger
x-ray units with greater diameters and lengths than those
30 discussed above could also be made and used in accordance
with the present invention.

To operate the x-ray unit 101 to prevent restenosis
in a vessel of the cardiovascular system, for example,
direct current having a voltage of between about 10-30 Kv,
35 can be applied to the central conductor 120. The external
conductor is connected to ground. Electrons emitted from
the cathode 110 due to a field emission effect impact the
anode 112, causing the emission of x-ray radiation of about
8-10 KeV, as discussed above. The radiation is primarily

1 emitted radially, to the vessel wall. About 10-30 Kv is
preferred for use in the prevention of restenosis. Higher
voltages will cause the emission of x-ray radiation of
higher energy which can penetrate too deeply into the vessel
5 wall, damaging cardiac tissue. Higher voltages may be used
for other applications.

Voltages at the higher end of the 10-30 Kv range are
preferred because the use of higher voltages enables the
generation of the same amount of radiation with less current
10 than the use of a lower voltages, and is therefore more
efficient. Higher voltages also enable the generation of x-
ray radiation of higher power. Higher power, however, can
cause the generation of more heat, which can damage the
tissue of a vessel wall. In this embodiment, most of the
15 heat is generated at the anode 110 positioned at the center
of the x-ray unit, as far from the vessel wall as possible.

Higher voltage also increases the risk of electrical
flashover at the anode and cathode triple junctions. As
discussed above, the anode 112 and cathode 110 are
20 preferably configured to minimize the risk of flashover.

Bulk electrical breakdown is also a risk with
increased voltages. Pyrolytic boron nitride has a high
dielectric strength, enabling the x-ray unit of the catheter
to tolerate the voltages used in this application without
25 bulk electrical breakdown. The dielectric strength of
pyrolytic boron nitride is 200-600 KV/mm.

Pyrolytic boron nitride is also particularly
preferred as the insulator 108 because it is highly
transparent to soft x-rays and can therefore be efficiently
30 used as an x-ray window. The coefficient of linear
absorption of boron nitride at about 8 Kev, the average
energy of the emitted radiation, is 1.0 mm^{-1} . About 8-10
KeV is the preferred energy level of x-ray radiation in the
treatment of restenosis, as discussed above. Transmission
35 of radiation through pyrolytic boron nitride with a
thickness of about 0.3 mm is about 70%. This enables
irradiation of tissue at a rate of at least about 1 gray per
minute. Preferably, about 10-30 grays per minute of
radiation at about 8-10 KeV are provided, enabling delivery

1 of an effective amount of radiation to prevent restenosis to
a lesion about 5 mm long in about 1 minute. It is believed
that x-ray radiation can be delivered at a rate of up to
about 100 grays per minute with the x-ray unit of this
5 embodiment. A lesion 1-2 cm long can be treated in about 2-
5 minutes by progressively repositioning the x-ray unit to
irradiate additional portions of the lesion.

Positive electrical pulses with a peak voltage of
between about 15-30 Kv and 2-100 nanoseconds long can also
10 be applied to the central conductor 120 of the coaxial cable
104 at a rate of between about 1-50 KHz. The high voltage
pulses cause field emission. The pulses can further cause a
vacuum electrical breakdown, causing electrons to flow from
the cathode 110 to the anode 112 through a plasma of
15 vaporized cathode and anode material between the cathode 110
and the anode 112.

The anode 114 is preferably attached to the
insulator 108 of pyrolytic boron nitride during formation of
the insulator 108 by chemical vapor deposition (CVD).
20 During CVD, the deposited boron nitride chemically bonds to
the anode material, forming a strong, vacuum tight seal.
The seal formed by CVD has higher voltage hold-off because
it does not have voids which can locally enhance the
electrical field and cause electrical flashover.

25 A mandrel 250 for use in manufacturing the x-ray
unit 102 by CVD is shown in Fig. 6. The mandrel 250 is
preferably graphite. A cavity 252 is provided in the
mandrel 250 for receiving the anode 114. The anode 114 is
secured in an anode holder 254 of boron nitride, for
30 example. The mandrel 250 includes a shoulder 254 for
supporting the metal ring 130. The metal ring 210 is held
in place by a cylindrical ring holder 256, also of boron
nitride, for example, which is supported by a mandrel holder
258 of graphite, for example.

35 The assembly of Fig. 6 is placed in a CVD reactor
for the deposition of boron nitride by CVD, as is known in
the art. Chemical vapor deposition of boron nitride is

1 described, for example, in Matsuda, et al., "Synthesis and
Structure of Chemically Vapour-Deposited Boron Nitride,"
Journal of Materials Science 21 (1986) pp. 649-658; and
5 Pouch, John J., et al. "Synthesis Properties of Boron
Nitride," Materials Science Forum, Volumes 54 and 55 (1990)
pp. 141-152, for example, which are incorporated by
reference, herein. The boron nitride is deposited on the
hot surface of the assembly, crystallizing into a hexagonal
10 structure. CVD of pyrolytic boron nitride can be performed
by CVD Products Incorporated, of Hudson, New Hampshire, for
example.

It may be advantageous to deposit and impregnate
boron onto the surface of the graphite mandrel 250 and
tungsten anode 114 prior to depositing the boron nitride.
15 To increase the chemical stability of the anode 114 during
the deposition procedure, the tungsten could be coated with
a layer of platinum about 1 micron thick.

After completion of the CVD process, the mandrel 250
is removed from the assembly by oxidation of the graphite,
20 also as known in the art.

The cathode 110 is then vacuum brazed to the metal
ring 130 with brazing materials, which are discussed below,
sealing the chamber. Vacuum brazing is also known in the
art and can be provided by Koral Labs., Minneapolis, St.
25 Paul, for example. The sealed chamber is then covered with
the conductive coating 124 by metal vapor deposition, for
example.

Such a process can be used for mass production of
large numbers of assemblies.

30 A fourth embodiment of an x-ray unit 300 in
accordance with the present invention is shown in Fig. 7.
The x-ray unit 300 comprises a vacuum chamber 302 defined by
an insulator 304, preferably of pyrolytic boron nitride, a
cathode 306, and an anode 308. The anode 308 is preferably
35 tungsten.

The cathode 306 may be graphite, titanium carbide,
graphite coated with titanium carbide or stainless steel,
for example. Graphite coated with titanium carbide is
preferred. A coating of several microns may be used.

1 Titanium coating can be provided by Lanxide Coated Products,
Inc., Newark, Delaware, for example. The cathode 306
preferably includes an annular protrusion 306c for creating
a cavity for containing the brazing material 316. The
5 cathode 306 may also include a protrusion 306a directed
towards the anode 308, as in the embodiment of Fig. 5.

The insulator 304 comprises a cylindrical wall 304a
with an inclined depending wall 310 and a cylindrical wall
314 preferably parallel to the cylindrical wall 304a. The
10 depending wall 310 is preferably angled towards the interior
of the vacuum chamber 302. The cylindrical wall 314 defines
a sleeve for receiving a depending portion 318 of the anode
308. The anode 308 is coupled to the cylindrical wall 314
through a brazing alloy 312. The cathode 306 is coupled to
15 the open end 314 of the insulator 304 through a brazing
alloy 316, as well.

The depending portion 318 of the anode 308
preferably includes a slot 320 for receiving the central
conductor 322 of a coaxial cable 324. The cathode 306 is
20 coupled to the external conductor 326 of the coaxial cable
324 through a conductive layer 325, as in the embodiment of
Fig. 5. A biocompatible coating is also provided over the
coaxial cable 324, conductive layer 325 and cathode 306. A
lubricious coating (not shown) may be provided, as well.

25 Preformed pyrolytic boron nitride of the desired
sizes and shapes is available from CVD Products,
Incorporated, for example.

Appropriate brazing alloys for coupling pyrolytic
boron nitride to the tungsten anode 308 include Incusil-15
30 ABA and Incusil-ABA, for example, available from GTE
Products Corporation, WESTGO Division, Belmont, C.A.
("WESTGO"). Incusil-15 ABA comprises 14.5% indium, 1.25%
titanium, 23.5% copper and 60.75% silver. Incusil-ABA
comprises 12.5% indium, 1.25% titanium, 27.5% copper and 59%
35 silver. The brazing temperatures for both alloys is about
750°C. The brazing material can be in the form of a
cylindrical ring placed within the sleeve formed by the
cylindrical wall 314 in Fig. 7. The brazing material
spreads into the vertical region between the anode 308 and

1 wall 314 during the brazing process. These alloys can also
be used to braze the cathode 110 to the metal ring 130 in
the embodiment of Fig. 5.

5 Appropriate brazing alloys for coupling a cathode
308 of graphite or graphite coated with titanium carbide to
pyrolytic boron nitride include Cusin-1 ABA and Cusil-ABA,
also available from WESTGO. Cusin-1 ABA comprises 34.25%
copper, 1.75% titanium, 1.0% tin and 63% silver. Cusil-ABA
comprises 63% silver, 35.25% copper and 1.75% titanium. The
10 brazing temperatures for both alloys is about 850°C. The
brazing is also conducted in a vacuum of about 10^{-5} Torr or
better. Because it requires a higher brazing temperature,
the graphite cathode 306 is coupled to the pyrolytic boron
nitride prior to the tungsten anode 308. The brazing
15 material can be in the form of a ring or it can be sputtered
onto the end of the pyrolytic boron nitride prior to vacuum
brazing.

Instead of a cathode of graphite, the cathode can be
PLZT or other such ferroelectric material, as discussed
20 above. As above, the use of ferroelectric material requires
the use of voltage pulses. In Fig. 8, a fifth embodiment of
the present invention is shown, comprising a ferroelectric
cathode 130 supported by a conductive cap 132. The
conductive cap 132 is coupled to the outer conductor 116 of
25 the coaxial cable 114 by a conductive layer 118, as above.
The remainder of the x-ray catheter 150 is the same as the
embodiment of Fig. 5. Graphite is preferred as the
conducting material because it has a low absorption
coefficient for x-ray, enabling transmission through the
30 distal end of the x-ray unit.

It is preferable to center the x-ray unit within the
vessel or lumen, to provide a uniform distribution of x-ray
radiation around the circumference of the vessel wall. Fig.
9 is a side view of an x-ray catheter 400 in accordance with
35 the present invention, with a centering device comprising a
plastic sleeve 402 with a plurality of resilient polymeric
solid arms 404 depending from it at an angle. The sleeve
402 can be coupled to the outer, biocompatible layer of the
coaxial cable 406 proximal to the x-ray unit 408 by adhesive

1 or thermal bonding, for example. The distal ends of the
arms 404 can optionally extend beyond the distal end of the
x-ray unit 408. The arms 404 bear against the vessel wall
410, centering the x-ray unit 408 within a vessel or lumen
5 of the body.

A sheath 412 is preferably provided over the coaxial
cable 406 for compressing the arms 404 during advancement of
the x-ray unit 408 to the intended site. When the x-ray
unit 408 is properly positioned, the sheath 412 is
10 retracted, releasing the arms 404. Radiopaque bands 414 of
gold or tantalum, for example, are preferably provided on
the coaxial cable 406 and the sheath 412 to assist in
tracking of the x-ray catheter 400 on a fluoroscope during a
procedure. The bands 414 are preferably positioned on the
15 coaxial cable 406 and the sheath 412 such that when the
sheath 412 has been sufficiently retracted to release the
arms 404, the bands on the coaxial cable 406 and the sheath
412 are essentially aligned.

Fig. 10 is a partial, cross-sectional view of the x-
20 ray catheter 400 of Fig. 9, wherein the x-ray unit 408 is
within the sheath 412 and the arms 404 are compressed.
Saline or some other cooling agent can be delivered through
the space 416 between the sheath 412 and the coaxial cable
406, as well.

25 Alternatively, a compressible cage 418 can be
provided over the x-ray unit 408 as a centering device, as
shown in Fig. 11. The cage 418 comprises a plurality of
arms 420 with a first end 420a coupled to a first sleeve
portion 422 and a second end 420b coupled to a second sleeve
30 portion 324. The x-ray catheter unit 408 extends into and
lies within the region defined by the arms 418. The arms
408 can be compressed by the sheath 412, as in Fig. 14. The
second portion 424 can be coupled to the distal end of the
x-ray unit 308.

35 The material of the outer layer of the coaxial cable
406 and the material of the sheath 412 preferably comprise
materials which slide easily with respect to each other.
The outer layer of the coaxial cable 406 is preferably

1 coated with a lubricious material, such as silicone or a
hyaluronic coating, as well.

Releasable arms and cages, methods of their
manufacture and suitable materials are disclosed in U.S.S.N.
5 08/488,216, filed on June 7, 1995 and assigned to the .pa
assignee of the present inventor. U.S.S.N. 08/488,216 is
incorporated by reference, herein.

Another method of centering the x-ray unit is a
malecot device, as shown in Figs. 12-13. A sheath 450 of
10 plastic material is attached to the distal portion 454a of
an x-ray unit 454, which is shown in Fig. 12. The coaxial
cable 456 attached to the proximal end of the x-ray unit, is
also shown in phantom. A plurality of lateral slots 457 are
provided through portions of the sheath surrounding the x-
15 ray unit 454. Four equidistantly positioned slots 457 may
be provided around the circumference of the sheath 450, two
of which are shown in Fig. 12. The length of the slots 457
depends on the diameter of the vessel at the intended site
and the diameter of the sheath 450, and should be sufficient
20 to enable the buckled portion of the sheath 450 to bear
against the circumference of the vessel wall. When the x-
ray unit 454 is adjacent the intended site, the sheath 450
is advanced, causing a portion 458 of the sheath 450 between
the slots 457 to buckle outward, as shown in Fig. 13. The
25 sheath 450 is advanced a sufficient distance for the portion
458 to buckle sufficiently to bear against the vessel wall,
centering the x-ray unit 454. The distal tip 460 of the
catheter may be of a soft, resilient material such as ultra
low density polyethylene or nylon, for example, as is known
30 in the art. Any of the embodiments of the x-ray catheter
can be provided with a soft tip.

The x-ray unit could also be placed within an
expandable balloon.

The x-ray catheters of the embodiments of Figs. 5, 7
35 and 8 can be conveyed to the site of the dilatation
procedure through an exchange tube after the dilatation
catheter is removed. The exchange tube can be advanced to
the intended site over the same guide wire used in the
dilatation procedure. After the exchange tube is properly

1 positioned, the x-ray catheters of Figs. 5, 7 and 8 can be advanced through the exchange tube, to the intended site.

The x-ray catheter of the present invention can also be advanced over the same guide wire used by the dilatation catheter after the dilatation catheter is removed, through a
5 guide catheter. Fig. 1 shows one such x-ray catheter 10. Fig. 14 is a cross-sectional view of another x-ray catheter 500 for use with a guide wire 502 in a rapid exchange configuration. The guide wire 502 enters the x-ray unit 504
10 through an opening 506 in the cylindrical wall of the unit 404, extends through the center of the unit 504 and a central passage 508 in a cathode 510, exiting through an opening at the distal end of the unit 504.

The cathode 510 of the x-ray unit 504 may be
15 graphite, for example. The anode can comprise a base 514 of tungsten, for example, with a plurality of rod-like protrusions 516 arranged concentrically about the base, within a vacuum cavity 518 defined by an insulator 520 and a cathode 510. The protrusions 516 extend toward the cathode
20 510. The insulator 520 is preferably of pyrolytic boron nitride. A tube 522 of insulative, vacuum tight material, may be provided through the vacuum chamber 518, providing a passage for the guide wire 502.

The base 514 of the anode has a depending portion
25 514a, preferably coupled to the central electrode 417 of a coaxial cable 518. A conductive layer is provided over the outer walls of the insulator 520, to couple the cathode 510 to the outer electrode of the coaxial cable 518, as described in the embodiments, above.

30 Fig. 15 is a side view of another embodiment of a rapid exchange x-ray catheter 600 in accordance with the present invention, wherein a portion of the catheter shaft 602 is shown in cross-section. Here, a lumen 601 is provided in the catheter shaft 602 with an entrance port 603
35 and an exit port 604 proximal to the x-ray unit 605. A guide wire 606 enters the lumen 601 through a port 603 and exits through a port 604. The x-ray catheter 600 can be tracked along the guide wire 606 to the intended site in a lumen or vessel, through the lumen 601. The distance

1 between the entrance port 602 and the exit port 604 can be
about 10-20 cm, for example. Other lumens (not shown) can
be provided for a coaxial cable or wires to couple the x-ray
unit 605 to the high voltage generator 32 shown in Fig. 1,
5 for example.

Such a catheter shaft 602 can be formed in a multi-
lumen extrusion process, as is known in the art, wherein the
lumens extend longitudinally through the catheter shaft 602.
The portions of the lumen distal and proximal to the
10 intended locations of the exit port 604 and entrance port
602 can be closed, as is known in the art. The ports 603,
604 can then be made through the catheter shaft by a laser,
for example.

While the above embodiments are described with
15 respect to applying x-ray radiation to the site of an
angioplasty procedure, the present invention can be used to
apply radiation within the cardiovascular system for other
purposes, or to other vessels, lumens, or cavities in the
body, wherever the application of radiation would be useful.

20 The various embodiments set forth above are for the
purpose of illustration. It will be appreciated by those
skilled in the art that various changes and modifications
may be made to these embodiments without departing from the
spirit and scope of the invention as defined by the claims,
25 below.

30

35

1 We claim:

1. A catheter for emitting x-ray radiation comprising:

5 a flexible catheter shaft having a distal end;
an x-ray unit coupled to the distal end, wherein the x-ray unit comprises an anode, a cathode and an insulator, wherein the anode and cathode are coupled to the insulator to define a vacuum chamber.

2. The catheter of claim 1, wherein the cathode is
10 a field emission cathode.

3. The catheter of claim 1, wherein the catheter shaft comprises a coaxial cable.

4. The catheter of claim 1, wherein the insulator is chosen from the group consisting of beryllium oxide,
15 aluminum oxide, or pyrolytic boron nitride.

5. The catheter of claim 1, wherein the cathode and the anode are coupled to a voltage generator.

6. The catheter of claim 1, further comprising a guide wire lumen.

20 7. The catheter of claim 6, wherein the guide wire lumen extends partially through the catheter shaft.

8. The catheter of claim 6, wherein the guide wire lumen extends partially through the x-ray unit.

9. The catheter of claim 1, further comprising a
25 means for centering the x-ray unit within a lumen.

10. The catheter of claim 1, wherein the cathode is a ferroelectric material.

11. An x-ray catheter comprising:

30 a flexible catheter shaft for being advanced through lumens of the vascular system, the catheter shaft having a distal end;
an x-ray unit coupled to the distal end, the x-ray unit comprising an anode, a cathode and an insulator, wherein the anode and cathode are coupled to the insulator
35 to define a vacuum chamber.

12. The catheter of claim 11, wherein the insulator comprises pyrolytic boron nitride.

- 1 13. The catheter of claim 11, wherein the anode comprises tungsten or platinum and the cathode comprises graphite.
- 5 14. The catheter of claim 11, wherein the cathode is a field emission cathode.
15. The catheter of claim 12, wherein the cathode and anode are coupled to a voltage generator.
- 10 16. The catheter of claim 15, wherein the catheter shaft comprises a coaxial cable coupling the anode and cathode to the voltage generator.
17. The catheter of claim 16, further comprising means for centering the x-ray unit within a lumen.
18. A catheter for the emission of x-ray radiation comprising:
- 15 a flexible catheter shaft having a distal end;
 an x-ray generating unit coupled to the distal end, the x-ray generating unit comprising an anode, a cathode and an insulator, wherein the anode and cathode are coupled to the insulator to define a vacuum chamber, and
- 20 wherein the x-ray generating unit has a diameter less than about 4 mm.
19. The catheter of claim 18, wherein the x-ray generating unit has a diameter of about 1 mm.
- 25 20. The catheter of claim 19, wherein the x-ray generating unit has a length of about 7 mm.
21. The catheter of claim 18, wherein the x-ray generating unit has a length less than about 15 mm.
22. The catheter of claim 18, wherein the insulator comprises pyrolytic boron nitride.
- 30 23. An x-ray catheter for use in irradiating the wall of a lumen comprising:
- a flexible catheter shaft having a distal end;
 an x-ray generating unit; and
 means for centering the x-ray generating unit
- 35 within the lumen.
24. A method for preventing restenosis of a lumen comprising:
- (a) advancing an x-ray catheter through a lumen to a first location adjacent an intended site of the

1 lumen, wherein the x-ray catheter comprises a flexible
catheter shaft with a distal end and an x-ray generating
unit coupled to the distal end, the x-ray generating unit
comprising an anode, a cathode and an insulator, wherein the
5 anode and cathode are coupled to the insulator to define a
vacuum chamber;

(b) causing the emission of an effective dose
of x-ray radiation to prevent restenosis; and

(c) removing the catheter.

10 25. The method of claim 24, wherein step (b)
comprises causing the emission of radiation within a
particular energy range to achieve a particular depth of
penetration.

15 26. The method of claim 24, wherein the causing
step (b) further comprises applying a predetermined voltage
between the anode and the cathode to achieve the particular
depth penetration.

27. The method of claim 24, further comprising
irradiating tissue at a rate of 1-100 grays per minute.

20 28. The method of claim 27, wherein the irradiating
step is conducted for about 1 minute.

29. The method of claim 24, wherein step (b)
comprises causing the emission of x-rays having an energy of
about 8-10 KeV.

25 30. The method of claim 24, further comprising
centering the x-ray unit within the lumen prior to the step
(b).

31. The method of claim 24, wherein the advancing
step comprises advancing the x-ray catheter through a lumen
30 of the vascular system through an exchange tube.

32. The method of claim 24, wherein the advancing
step comprises advancing the x-ray catheter through a lumen
of the vascular system over a guide wire and through a guide
catheter.

35 33. The method of claim 32, wherein a portion of
the x-ray catheter is advanced over the guide wire.

34. The method of claim 24, further comprising
positioning the x-ray unit at a second location and causing
the emission of x-ray radiation at the second location.

- 1 35. The method of claim 24, further comprising
positioning the x-ray unit at a plurality of locations and
causing the emission of x-ray radiation at each of the
plurality of locations.
- 5 36. The method of claim 24, further comprising
conducting an angioplasty procedure prior to step (a),
wherein the intended site of step (a) is the site of the
angioplasty procedure.
- 10 37. A method for providing x-ray radiation
treatment comprising:
 advancing an x-ray catheter through a lumen to
an intended site, wherein the x-ray unit comprises a
flexible catheter shaft with a distal end and an x-ray
generating unit coupled to the distal end, the x-ray
15 generating unit comprising an anode, a cathode and an
insulator, wherein the anode and cathode are coupled to the
insulator to define a vacuum chamber;
 causing the emission of an effective dose of x-
ray radiation; and
20 removing the catheter.
38. The catheter of claim 2, wherein the cathode is
chosen from the group consisting of graphite, titanium
carbide, carbides, metals, and graphite coated with titanium
carbide.
- 25 39. The catheter of claim 1, further comprising a
guide wire lumen extending through the catheter shaft.
40. The catheter of claim 2, wherein the cathode
comprises silicon and the x-ray unit further comprises a
grid proximate the cathode.
- 30 41. The catheter of claim 2, wherein the cathode
comprises silicon needles.
42. The catheter of claim 11, wherein the x-ray
unit irradiates tissue at a rate of at least about 1 gray
per minute.
- 35 43. The catheter of claim 1, wherein the anode is
coupled to a wall of the insulator, wherein the wall is
tapered towards the anode.
44. The catheter of claim 3, wherein:

1 the coaxial cable comprises an outer conductor
and a central conductor;

 the insulator has a tubular portion with
proximal and distal ends, the coaxial cable being coupled to
5 the proximal end, the anode being coupled to the proximal
end and to the central conductor of the coaxial cable, and
the cathode being coupled to the distal end;

 the catheter further comprises a conductive
surface surrounding the tubular insulator, coupling the
10 cathode to the outer conductor of the coaxial cable; and

 the insulator and cathode define an annular
region proximate the coupling between the cathode and the
insulator, the annular region being screened from an
electrical field generated between the anode and the cathode
15 by the conductive surface and a portion of the cathode.

 45. The catheter of claim 44, wherein the insulator
comprises a wall depending from the proximal end of the
tubular portion, the wall being angled toward the anode and
the vacuum chamber.

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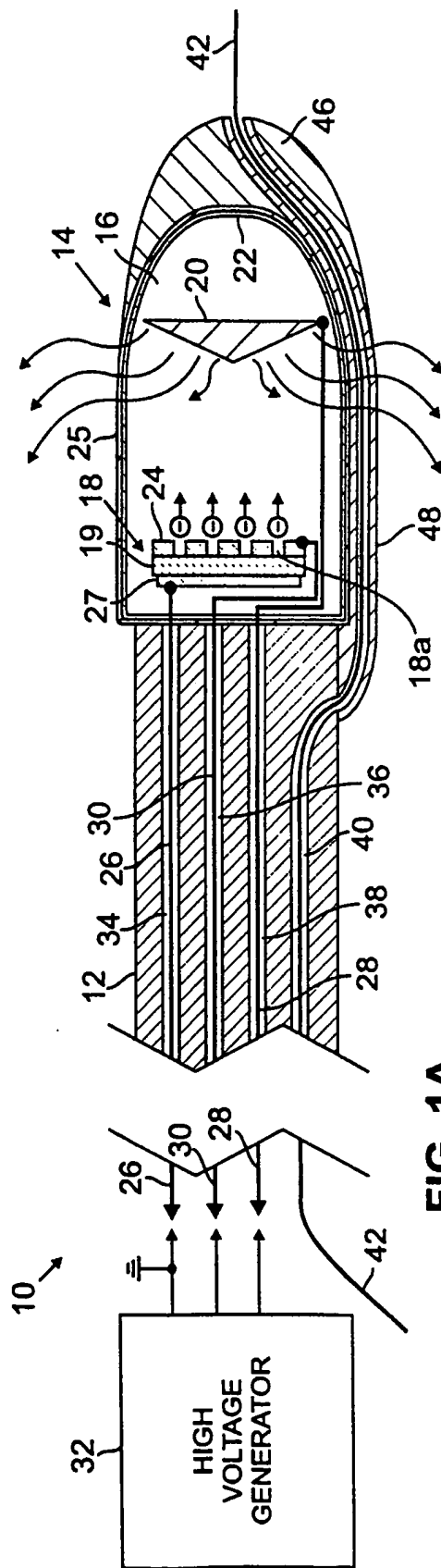


FIG. 1A

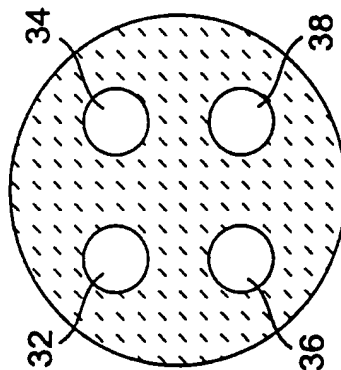
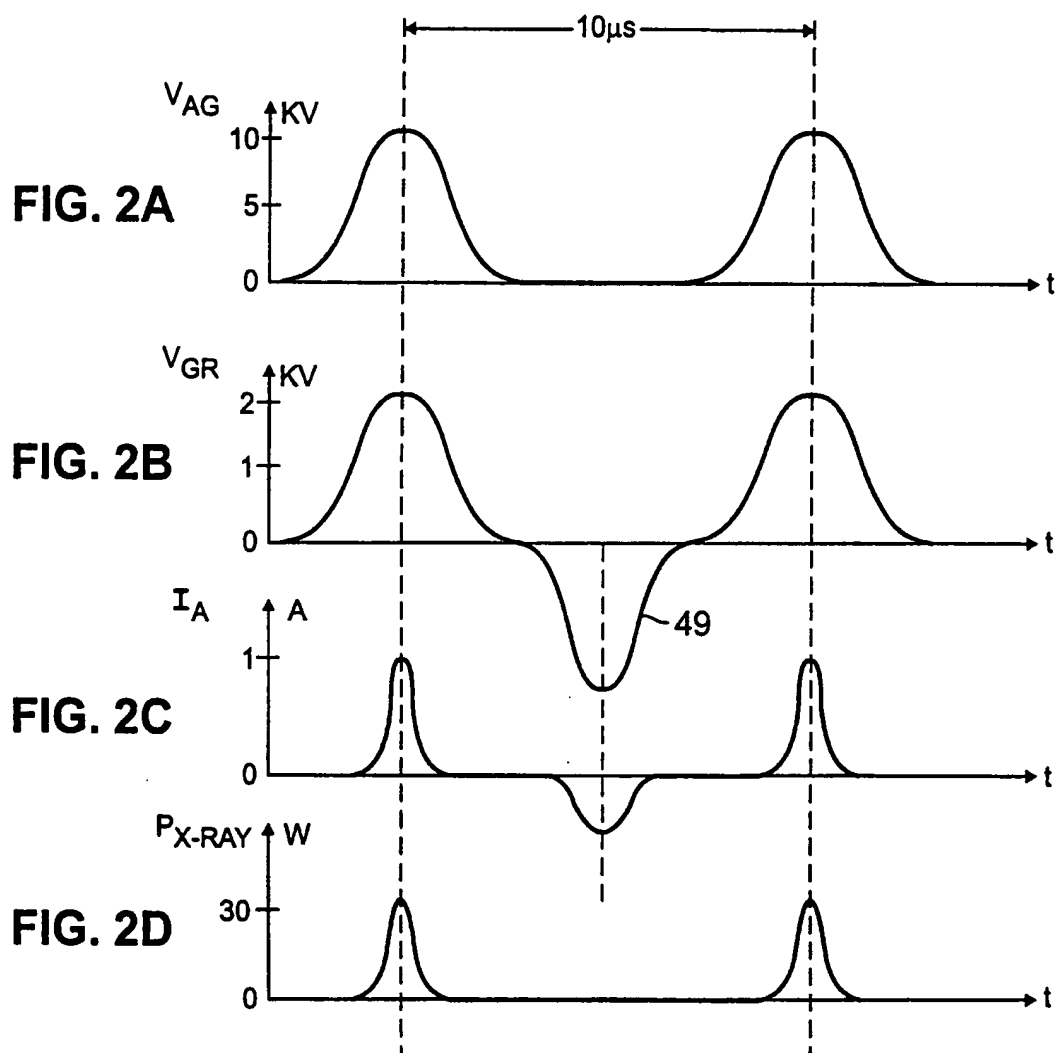
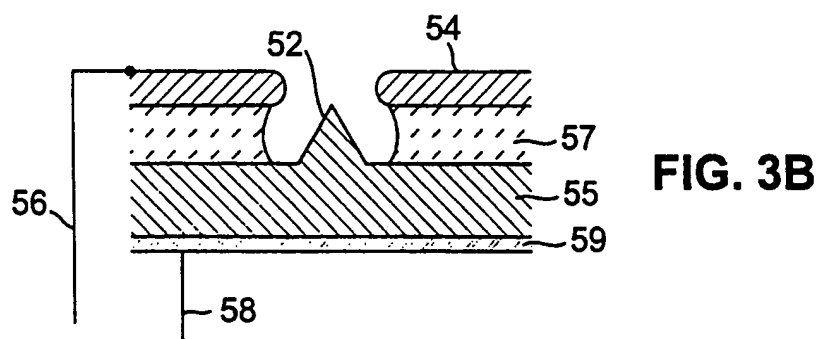
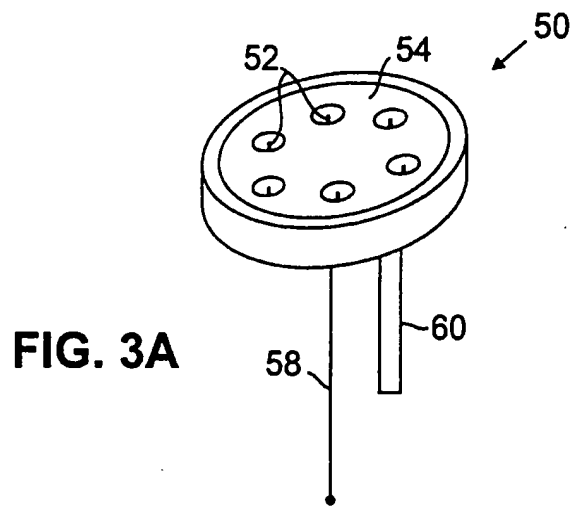


FIG. 1B

2/11



3/11



4/11

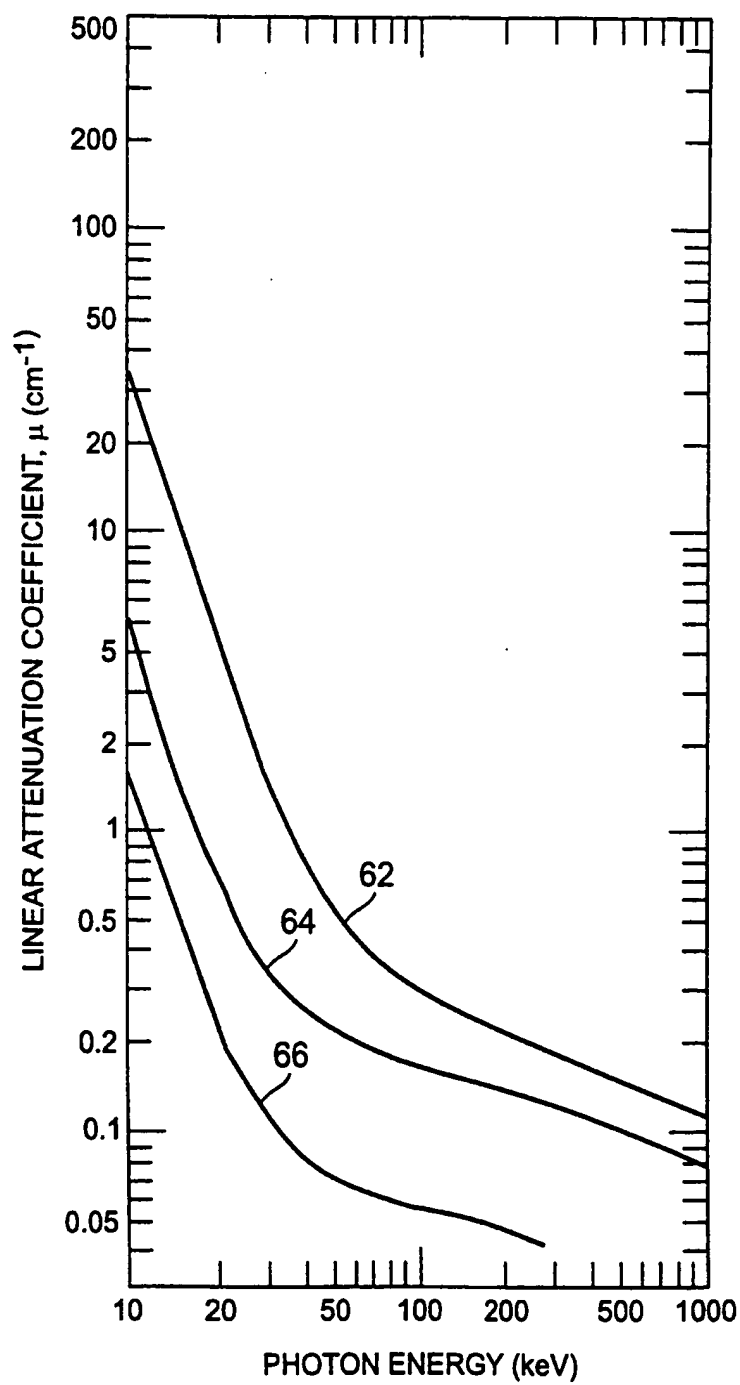


FIG. 4

5/11

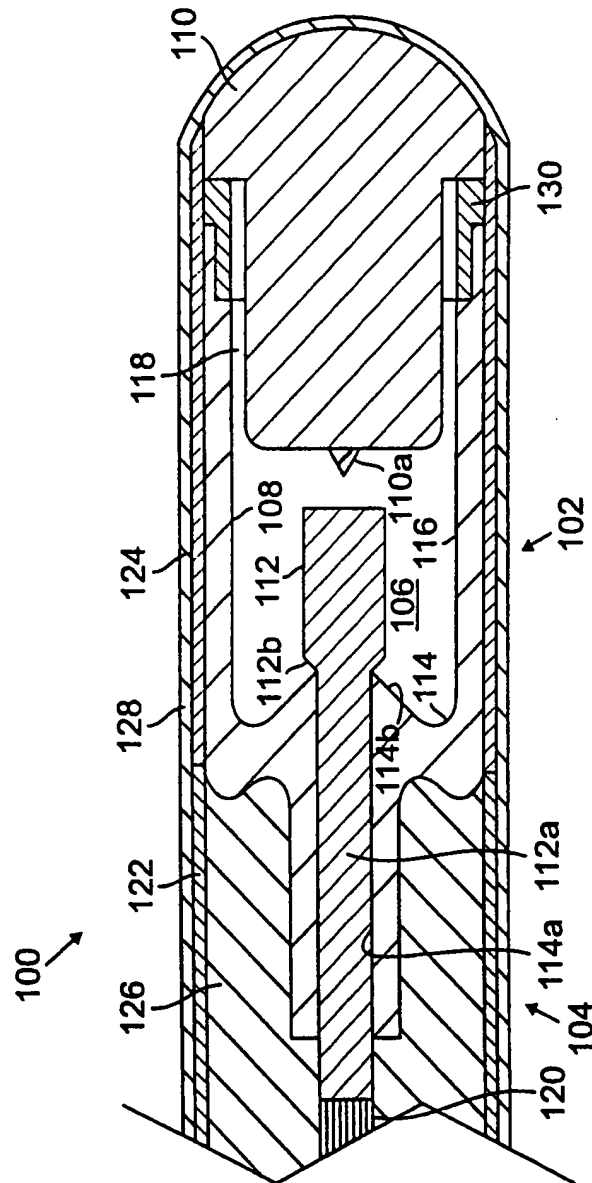
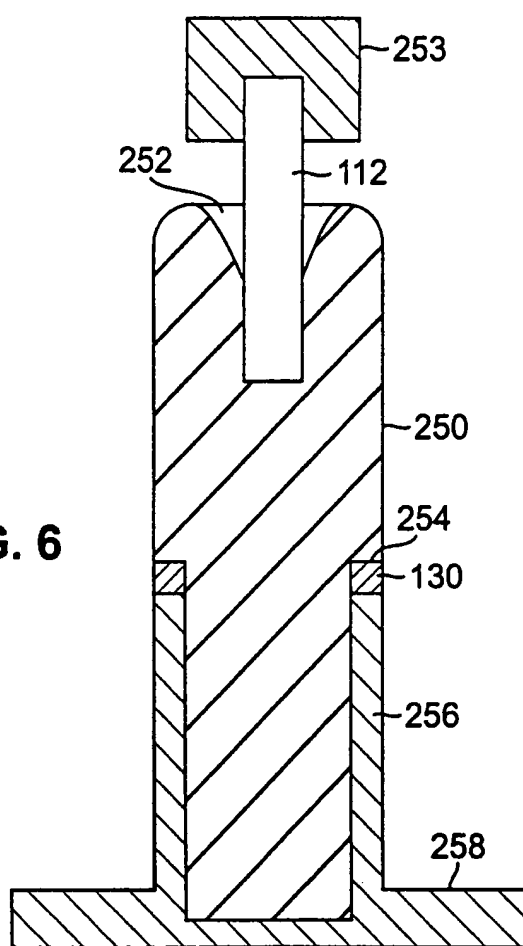


FIG. 5

6/11

FIG. 6



7/11

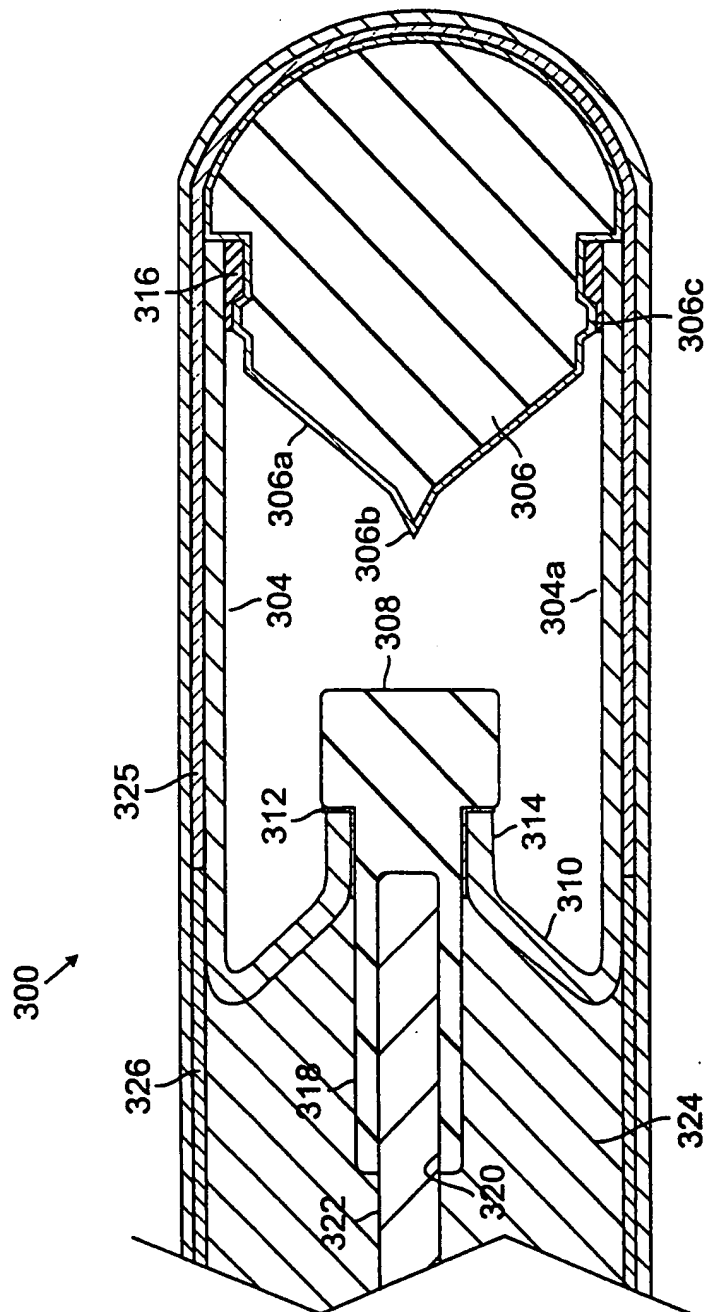


FIG. 7

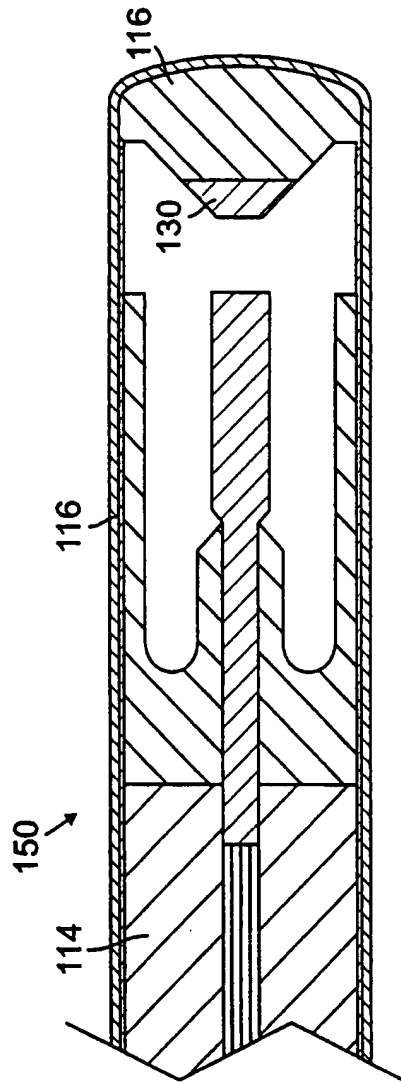


FIG. 8

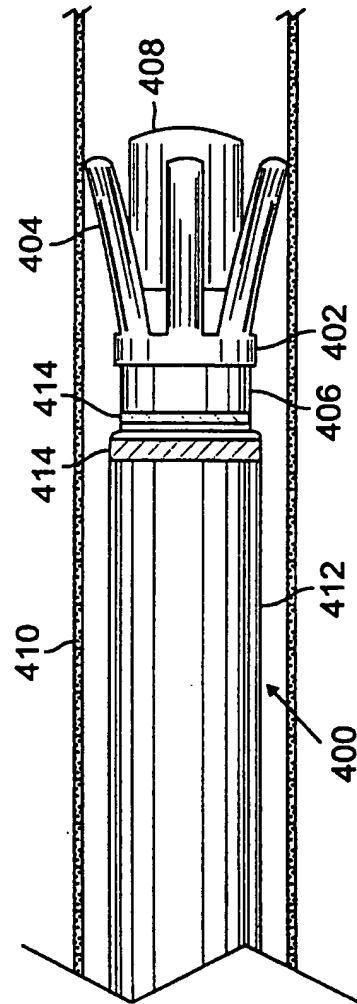


FIG. 9

9/11

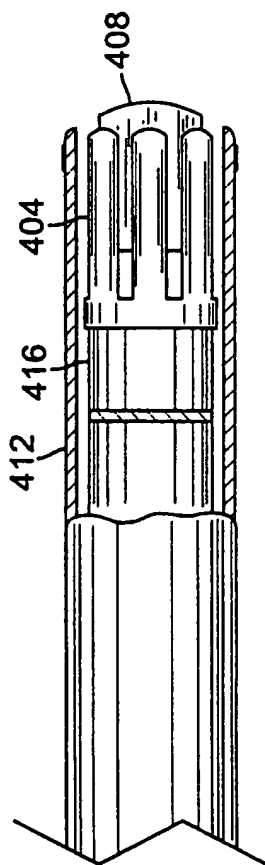


FIG. 10

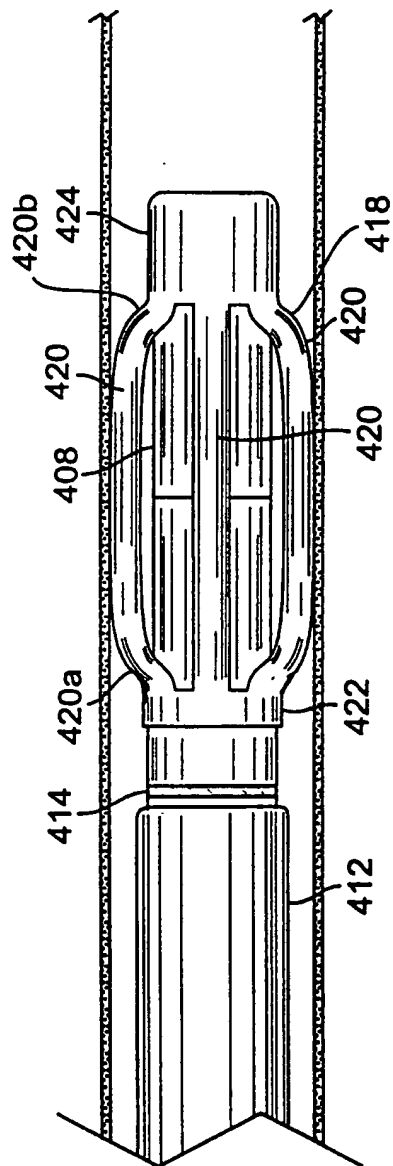


FIG. 11

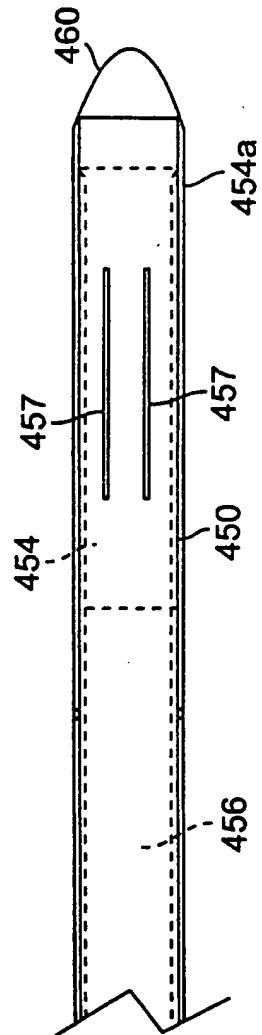


FIG. 12

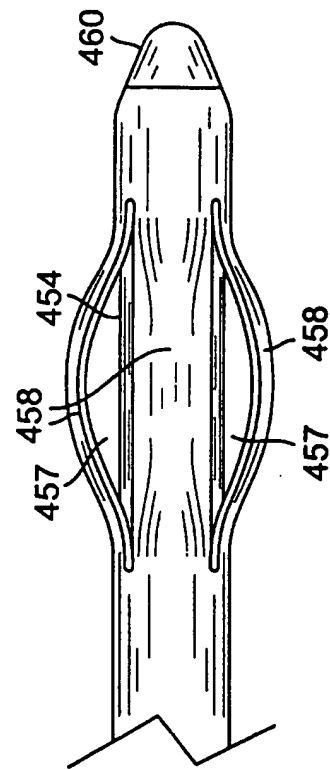


FIG. 13

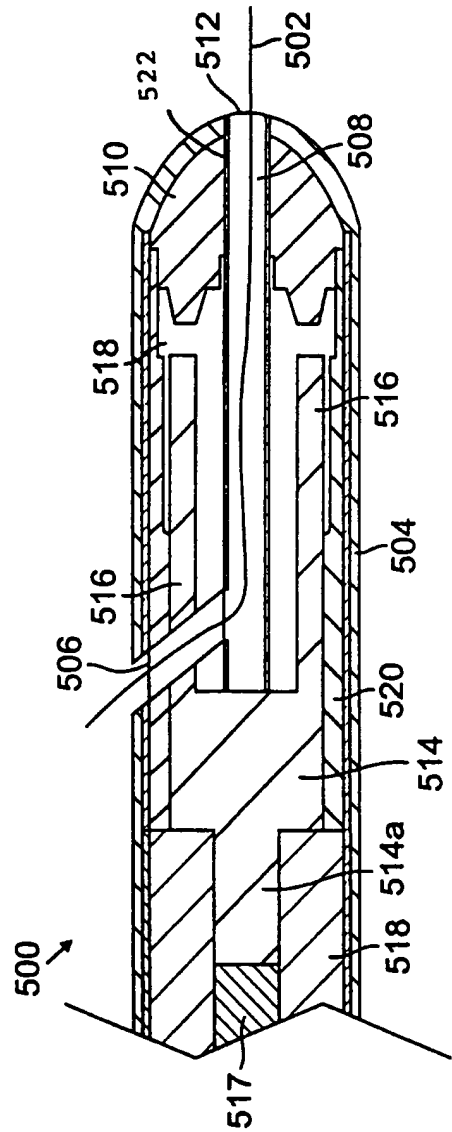


FIG. 14

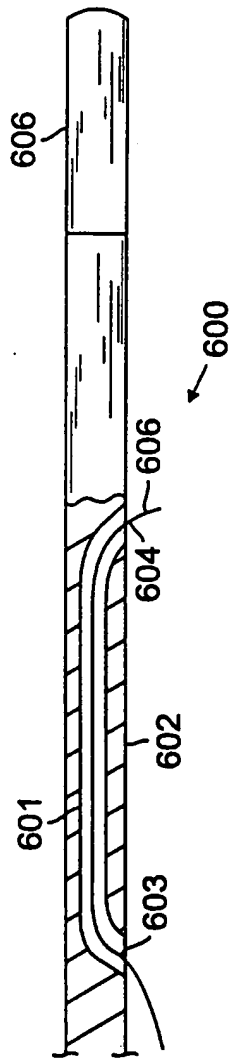


FIG. 15

INTERNATIONAL SEARCH REPORT

Insert international application No.
PCT/US96/13629**A. CLASSIFICATION OF SUBJECT MATTER**

IPC(6) :A61B 6/00; A61N 5/00

US CL :128/653.1, 659

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 128/653.1, 659; 600/109; 606/1, 2, 7, 32; 607/804

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

APS

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 5,199,939 A (DAKE et al) 06 April 1993, entire document.	1
A,P	US 5,503,613 A (WEINBERGER) 02 APRIL 1996.	1
A	SU 814-331 A (A MED ONOLOGY RES) 23 March 1981.	1-45
A	DT 2054 738 A (LOMMATZSCH et al) 10 May 1972.	1-45



Further documents are listed in the continuation of Box C.



See patent family annex.

* Special categories of cited documents:	"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"A" document defining the general state of the art which is not considered to be of particular relevance	"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
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"O" document referring to an oral disclosure, use, exhibition or other means	
"P" document published prior to the international filing date but later than the priority date claimed	

Date of the actual completion of the international search

11 NOVEMBER 1996

Date of mailing of the international search report

15 NOV 1996

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